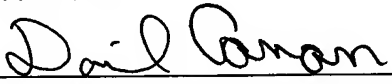PATENT APPLICATION COVER SHEET
Attorney Docket No. 0321.68297

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9/17/2003
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SCINTILLATION CRYSTAL DETECTION ARRAYS
FOR RADIATION IMAGING DEVICES

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SCINTILLATION CRYSTAL DETECTION ARRAYS FOR RADIATION IMAGING DEVICES

REFERENCE TO RELATED APPLICATION AND PRIORITY CLAIM

5   This application is related to pending provisional application serial number 60/411,613 filed on September 18, 2002, and claims priority from that provisional application under 35 U.S.C. §119.

FIELD OF THE INVENTION

10   The field of the invention is high resolution radiation imaging. An exemplary application of the invention is a Positron Emission Tomography (PET) device.

BACKGROUND OF THE INVENTION

     Scintillation crystals are commonly used in non-invasive medical diagnostic 15 techniques that utilize radiation emitting materials. These crystals are noted for their ability to emit pulses of visible light when ionizing radiation, such as gamma radiation, passes therethrough and interacts with atomic nuclei in the crystal. The pulses of emitted light (photons) are then detected by a photodetector device such as a photomultiplier tube (PMT) or a semiconductor photodiode (SPD). The 20 effectiveness of the detector in diagnostic procedures depends on the ability to see and quantify the crystal light flashes with high spatial, spectral, and temporal precision. This, in turn, is dependent on brightness and rapidity of the generated and recorded flash, which are functions of the type and geometry of the scintillation crystal.

25   Position emission tomography (PET) is an example radiation imaging technology that provides in-vivo, functional information about the molecular biochemistry of a given radio-labeled compound (a.k.a. tracer) introduced into a live

subject. The radio-label is a positron emitter. The tomographic imaging is possible through detection and localization of the many associated highly energetic photons emitted. The sensor is typically an array of scintillation crystals. There are essentially three stages of the photon sensing. The entering photon is first absorbed

5   in the scintillation crystal. The crystal gives off a flash of light. This light is collected by a photodetector, which subsequently detects and converts the light into electric charge that is amplified. The result is a robust electric signal with an amplitude that represents the energy of the incoming photon, a location that indicates where the energetic photon came from within the imaging subject, and

10   time stamp that signifies when the event occurred. For high spatial resolution imaging, which will allow one to see very minute structures, PET relies on very accurate localization of the energetic photon emissions. This means the scintillation detector must have very fine position resolution of the entering photons. However, to efficiently absorb the incoming photons, the crystal must also be relatively thick.

15   Efficient absorption of incoming photons is important to allow for high count sensitivity, which translates into good image quality.

In typical commercial devices, there has been a compromise between detector spatial resolution, detection efficiency, and light collection into the photodetector. High light collection is important for creation of robust electronic detector signals

20   for high sensitivity. High light collection is also important for good energy resolution for good photon Compton scatter and random coincidence background rejection capabilities. Scatter and random photon coincidences produces loss of image contrast and should be rejected as much as possible. Efficient light collection is also important for good event timing resolution that further helps to reject random

25   coincidence background.

In typical commercial PET devices spatial resolution is determined by the dimensions of the individual scintillation crystals. Typical commercial nuclear

medicine cameras utilize a two-dimensional, discrete or pseudo-discrete array of long, narrow scintillation crystals, which are coupled at a small end to photomultipliers (PMTs) with the opposite small end directed toward the high energy photon source. Positional information is encoded in the crystals, as each

5   crystal indicates a unique X-Y position. The crystals are made long for high gamma-ray stopping power and narrow for high spatial resolution. An intermediate optical coupling medium is necessary in these designs at the scintillation crystal/PMT interface. Crystal surfaces in these designs are treated and coated with reflectors to preferentially direct light through internal reflections into the PMT

10   located at one small end of the scintillation crystal array. Detection sensitivity in the conventional discrete-crystal designs is hampered by coupling of the scintillation crystals to the PMTs. Another type of commercial device used in Nuclear Medicine utilizes a scintillation crystal sheet, with the radiation incident on the large surface of the crystal sheet. Detection by a PMT is on an opposite large surface of the crystal

15   sheet. This configuration is commonly known as a scintillation camera.

The typical commercial PET devices collect emitted light photons at the opposite end from the end at which the radiation of interest is received in the long, narrow crystals.. The front as well as the side crystal surfaces include reflective coatings to help alleviate loss of intensity in the scintillation light resulting from the

20   received radiation. However, for very fine crystals in high resolution PET systems only a small fraction of the scintillation light produced in the crystal reaches the photodetector. This light loss problem associated with standard photodetector readout at the end of the scintillating crystal worsens as the crystal is made narrower and longer or has untreated surfaces. This light loss problem together with the low

25   quantum efficiency of the PMT photocathode for detecting the scintillation light produced limits the count efficiency and signal-to-noise ratio of both the crystal decoding scheme used to position and time an annihilation photon event, and the

energy (spectral) resolution required to reduce gamma-ray scatter and random coincidence background rates. Good scatter and random photon reduction is an important factor for improving image contrast between true structures of interest and the background present in the resulting PET images.

5   An additional related problem associated with the conventional end readout is that the light collection efficiency depends on the location within the crystal where light was created and thus, where the radiation interacted. This factor further degrades the energy resolution. Also, there is roughly a 10-15% light loss at the interface between the crystal and PMT due to index of refraction mismatches,

10   further degrading the signal to noise ratio. In a particular commercial high resolution design, additional fiber optics are included between the crystals and the PMT, which further degrades the available light signal.

Extracting a high fraction of the available scintillation light from the ends of long and narrow crystals proves to be very difficult due to a poor aspect ratio for

15   light collection. The result is lower signal to noise ratios (S/N), relatively small pulse heights (reduced sensitivity), and inadequate energy resolution (reduced Compton scatter and random photon rejection capabilities). This low light extraction also contributes to non-optimized coincidence time and spatial resolution. To facilitate light collection, the crystal sides must be highly polished, which

20   significantly increases complexity and costs.

The state of the art was advanced by the invention described in U.S. Patent 6,114,703 (incorporated by reference herein) to Levin et al. The '703 patent provided an efficient method and devices for collection, and made the large surfaces of long and narrow scintillation crystals available for detection. The '703 patent

25   disclosed methods and devices that replaced the bulky and expensive PMTs by utilizing semiconductor photodetectors, applying such semiconductor photodiodes directly to surfaces of the scintillation crystals, including at least one large surface of

the scintillation crystal. The device of the '703 patent improved the amount of light measured from a scintillation event, while maintaining high spatial resolution offered by long and narrow scintillation crystals. The '703 patent also improved upon the single sheet style conventional devices that receive radiation perpendicular

5   to the large face of the crystal sheet by eliminating the coupling losses associated with the optical interfaces between the crystal and PMT and replacing the PMT of the conventional devices with directly deposited semiconductor photodiodes.

An overriding goal in radiation imaging is to obtain reconstructed images of very high spatial resolution. Spatial resolution improvements in reconstructed

10 images have come most often from reductions in the size and increases in the number of scintillation crystals. Detection sensitivity, though, is another limiting factor. The '703 patent was directed to improvements in the detection sensitivity. To maintain high detection sensitivity and good image quality, the challenges are to develop a finely pixellated scintillation crystal array with both high detection

15 efficiency and high light collection. High detection efficiency means the crystals must be relatively long, tightly packed, and cover a relatively large axial field-of-view (FOV).

A difficulty with designs having small scintillation crystals for high resolution is that manufacturing is a significant challenge. It is costly and complex

20 to handle many minute crystal elements and align them with corresponding photodetector elements. Slight misalignments might reduce light collection efficiency. A shortcoming with conventional crystal sheet devices for PET is that the sheet must be thin so that it produces a relatively narrow beam of light onto the photodetector plane. Thus, crystal sheet detectors that have been used in PET suffer

25 from low efficiency for stopping the high energy photons.

SUMMARY OF THE INVENTION

A radiation imaging device includes scintillation crystal sheets arranged in parallel to each other. Semiconductor photodetector positional detectors read light from large faces of the scintillation crystal sheets to detect interactions in the

5   scintillation crystal sheets and independently provide positional information concerning the interactions relative to at least one axis.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 is a schematic representation of a preferred embodiment PET device

10   with a scintillation crystal array receiving photon emissions on end faces of sheet scintillation crystals;

FIG. 2 is a schematic representation of a preferred embodiment scintillation crystal array;

FIG. 3 is a schematic representation of a preferred embodiment PET device

15   with a scintillation crystal array receiving photon emissions in an orthogonal direction with respect to sheet scintillation crystals in the crystal array;

FIG. 4 is an exploded partial schematic view of a preferred embodiment scintillation crystal detection array;

FIG. 5 is a schematic view of an avalanche photodetector line array usable in

20   embodiments of the invention;

FIG. 6 is an exploded partial schematic view of a preferred embodiment scintillation crystal detection array;

FIGs. 7A and 7B show preferred embodiment scintillation crystal detection array arranged as a ring, and an associated photodetector array component,

25   respectively;

FIG. 8 shows a preferred embodiment modular mosaic scintillation crystal detection area for imaging large area portions;

FIGs. 9A and 9B illustrate a preferred modular scintillation crystal array having modules arranged in multiple rings;

FIGs. 10A and 10B show simulated light distributions at a photodetector plane for a scintillation crystal sheet; and

5   FIGs. 11A and 11B show extracted FWHM values and a sample fit (for distribution 5 of FIG. 10B).

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

The invention relates to scintillation crystal arrays and imaging devices using

10   scintillation crystal arrays. A scintillation crystal array of the invention increases light collection, can obtain near complete light collection, and improves detection, count efficiency, spatial resolution, and analysis of penetrating radioactive emissions. A preferred application of the invention is Positron Emission Tomography (PET).

15   A preferred embodiment of the invention includes an array of scintillation crystal sheets arranged in a device such that radiation is incident upon small end faces of the sheets. Semiconductor photodiodes read light from large faces of crystal sheets. The semiconductor photodiodes may be pixellated, meaning that the semiconductor photodiodes provide both detection of photons generated in the

20   scintillation crystals and positional information about a detection, or may be one large pixel with positioning capability within that pixel   In another preferred embodiment of the invention, radiation is incident perpendicular to a large face of scintillation crystals. The semiconductor photodiodes are essentially transparent to the radiation, which accordingly may pass through the face of one of the scintillation

25   crystals of the array and into other ones of the scintillation crystals of the array. The semiconductor photodiodes are position sensitive.

Devices of the invention include high energy photon detectors or cameras for either detection, spectroscopic or imaging uses. In Nuclear Medicine, certain tissue within the body absorbs a particular radiopharmaceutical at a differential rate producing a distribution of the radioisotope. Counting the photon emissions with a

5   tomographic array of detectors enables the creation of cross-sectional images showing a spatial distribution of how the organs and tissue utilize a particular radiopharmaceutical. The present invention is especially well-suited for Nuclear Medicine Imaging (NMI), such as gamma-ray cameras (Single Photon Emission Computed Tomography, (SPECT) or Planar Imaging or Positron Emission

10   Tomography (PET). As an example, PET benefits from detecting with high precision the high energy photon activity of small radio-labeled structures within living subjects. Applications of high resolution PET include pharmaceutical and radio-pharmaceutical development, oncology, gene therapy and neuroscience.

Referring now to FIG. 1, a preferred embodiment PET device 10 is shown. A

15   subject radiation station 12 of the device is conventional, and is not illustrated in detail. The subject radiation portion 12 may be the same or similar to commercial PET devices, such as those manufactured by GE and Seimens. The subject radiation portion 12 of the PET device produces emissions 14 that are detected by a scintillation crystal detection array 16. The scintillation crystal detection array 16

20   includes a plurality of thin sheets 18 of scintillation crystals and intervening semiconductor photodetector positional detectors 20. Sheet, as used herein, refers to a crystal that is optically continuous over a plurality of positions that can be detected by one or more photodetector semiconductor positional detectors. The thin sheets 18 are either continuous sheets of scintillation crystal or discrete crystals in optical

25   contact with each other so that they share light. The semiconductor photodetector positional detectors are arranged to detect interactions producing light photons in the scintillation crystals 18, and to provide positional information concerning the

location of interaction. In the FIG. 1 embodiment, detection of interaction in a particular crystal sheet 18 by a corresponding semiconductor photodetector positional detector provides positional information relative to a first axis (perpendicular to the large faces of the crystal sheets 18) and the semiconductor

5   photodetector positional detectors themselves independently provide positional information relative to at least one second axis that is perpendicular to the first axis. Total detection information provided includes positional, temporal and spectral information.

The semiconductor photodetector positional detectors 20 comprise

10   semiconductor photodetectors, such as lines (see, e.g., FIGs. 4 and 5), coupled to the crystal sheets 18 with or without an intermediate optical coupling compound. The photodetectors may be formed directly on the crystal sheets by conventional fabrication techniques, e.g., crystal growth from solution or melt, chemical vapor deposition, or molecular beam epitaxy or simply epoxied or otherwise attached to

15   the sheets. Another possibility is to use a substrate 23, e.g. a thin ceramic, for structural support of the photodetectors. In the FIG. 1 embodiment, the scintillation crystal detection array 16 is arranged such that the emissions 14 are incident on a small crystal end face 24 of the crystal sheets 18. The emissions 14 are generally parallel to large faces 25 of the crystal sheets 18.

20   Referring now to FIG. 2, an exemplary embodiment of an array of 1mm thick crystal sheets 18. The crystal sheets 18, for example, dimensioned as 10x10x1 mm^3, are advantageous compared to long and narrow crystal fingers, in that it is significantly easier and less expensive to cut larger crystal pieces and assemble them into an array. The orientation of the FIG. 1 embodiment relative to that of incoming

25   photons (gamma radiation) is at the end faces 24 of the crystal sheets 10. Preferably, the semiconductor photodetectors are $\leq 300\mu m$ in thickness. Using crystal sheets 18 may permit a very thin array since it may be possible to completely remove the array

substrate and use the crystal instead as backing for mechanical strength to support the silicon wafer. Any type of position sensitive photodetectors with appropriate properties may be used. This includes line and square-pixel avalanche photodiode arrays, and position sensitive avalanche photodiodes. The position sensitive

5 avalanche photodiodes are preferred as photodetector pixellation or segmentation are not required, and positional information may be extracted from only four readout channels.

Another embodiment is shown in FIG. 3. The structure of the scintillation crystal detection array 16 is similar to that in FIG. 1, but in FIG. 3 incident gamma

10 radiation is received by a large crystal face 25 of one of the crystal sheets, i.e., generally orthogonal to the large faces of the crystal sheets 18. In this case, the semiconductor photodetector positional detectors 20 are required to independently provide positional information relative to two axes, namely an X and Y axis parallel to the large faces of the crystal sheets 18. The particular sheet 18 in which an

15 interaction is detected provides a Z axis depth determination.

The FIG. 3 embodiment allows higher detection sensitivity since incoming photons do not see any crystal gaps, unlike in FIG. 1 where some incoming photons are incident upon spaces, though possibly very small spaces, between the individual sheets 18 of the array 16. In the FIG. 3 embodiment, the orthogonal orientation also

20 eliminates the need to make the semiconductor photodetectors very thin. For the incident gamma radiation, silicon semiconductor photodetectors are substantially transparent, and accordingly do not adversely affect the penetration of gamma radiation photons into the crystal sheets 18.

As has been mentioned, the semiconductor photodetector positional detector

25 in the FIGs. 1 and 3 embodiments must provide position sensitivity. The position sensitivity may be achieved by different photodetector types, arrangements and techniques. Example semiconductor photodetectors that could be used include line

arrays of photodetectors, such as avalanche photodiode (APD) line arrays, or position sensitive photodetectors, such as position sensitive APDs (PSAPDs). PSAPDs have the advantage that the scintillation light pulse is positioned using only four APD signals. This would reduce the number of electronic channels in the

5   system. Preferred exemplary photodetector arrangements are shown in FIGs. 4 - 6.

FIG. 4 is an exploded partial view of a scintillation crystal sheet 18 and two planes 30, 32 of APD photodetectors 34. The planes 30, 32 form alternate planes of APD line arrays oriented in the x and y direction on either side of each crystal sheet 18 in a cross-grid type readout. The position of an incident photon causing a light

10   photon to be generated by the crystal 18 is detected by an APD 34 in each of the planes 30, 32 and a signal is provided on corresponding leads 38 to suitable processing. An interaction point in any crystal sheet plane could be determined by which cross-strip x and y elements were hit. Depth of interaction information is determined by which APD planes provide a signal. The same cross-grid style APD

15   planes can be used in the FIG. 1 embodiment. The cross-grid arrangement would provide information about the depth of interaction in the FIG. 1 embodiment.

The FIG. 1 embodiment may simply use APD line arrays that are parallel to the path of incident radiation, though the depth of interaction may or may not be provided. An example APD line array 40 is shown in FIG. 5, and includes a

20   plurality of APD line detectors 34 spaced apart from each other. Each provides a signal to a corresponding electronic lead 38. With the line array 40 replicated between each crystal sheet 18, X and Y positional information is provided. The resolution along one axis is determined by the pitch of the APD line detectors 34 and the resolution along another axis is determined by the location of the crystal layers

25   18. It is also possible to use APD line elements that are segmented along the length of each element as depicted in FIG. 5. The signals from the different segments 34a, 34b coarsely provide the interaction depth location. A segmented APD would

provide the X and Y positional information directly. Another possibility is a multiplexing scheme in which the elements 34 are all connected through resistive division so that only a few (e.g. 4) readout signals are required.

FIG. 6 illustrates an embodiment that uses a PSAPD 42 between each crystal

5   sheet 18. FIG. 6 indicates the orthogonal orientation of FIG. 3, but the orientation of FIG. 1 may also use a PSAPD for the semiconductor photodetector positional detectors 20. A PSAPD as shown in FIG. 6 is able to position a photon interaction 43 using only four PSAPD signals. When applied to either of the FIGs. 1 and 3 embodiments, both X and Y positional information and depth of interaction

10   information is provided automatically without any physical segmentation of photodetector elements.

FIGs. 7A and 7B show a preferred embodiment scintillation crystal detection array 16 arranged as a ring, with individual crystal sheets 18 receiving radiation at their small ends. Intervening photodetector positional detectors 20, for example the

15   APD line array of FIG. 7B detect light photons produced by interactions in the crystal sheets. The FIGs. 7A and 7B embodiment with a relatively wide APD line array, e.g., 4cm, provides relatively large axial coverage (4 cm). This axial field of view could cover a large fraction of the length of a small animal subject, e.g., a mouse, or a body portion, e.g., a hand, for high sensitivity imaging. Currently

20   silicon technology allows arrays as wide as 6 cm. If a longer axial dimension is desired in 7A, multiple rings of detectors may be pieced together.

FIG. 8 shows another scintillation crystal detection array, for imaging large area portions, e.g., human breast examinations. A large area may be covered in the example of FIG. 8 by tiling together multiple array modules. FIG. 8 represents four

25   scintillation crystal detection array modules 16a, 16b, 16c, and 16d. The modules can be arranged in a planar mosaic pattern as shown in FIG. 8 to image a large area. The FIG. 8 embodiment is a planar detector arrangement with emissions incident on

ends 24 of crystal sheets 18 as in the FIG. 1 embodiment. Modules 16n may also be arranged in multiple rings to form a cylinder to cover a large axial area, as shown in FIGs. 9A and 9B. This arrangement, with a large number of modules, may also be expanded to a sufficient diameter cylinder to image large subjects. Leads 38 may be

5   exposed, for example, on an outer circumference of the rings of modules 16n.

To study signal to noise ratio capability, light collection from a large face of a thin, 10x10x1 mm^3 LSO crystal sheet was studied. Point light sources were simulated at the center and edge of the large area of a crystal sheet. For these central and edge locations, three different source distances were simulated with respect to

10   the photodetector plane as shown: a top position corresponded to ~0.1 mm from the top face as shown, a middle position to the crystal center, and a bottom position to ~0.1 mm from the bottom plane where the photodetector is coupled. The total light collection from the large face was nearly complete independent of crystal surfaces and source position.

15   The simulated light distributions seen at the photodetector plane are shown in FIGs. 10A and 10B. For source positions very close to the photodetector plane, the exiting light distribution is extremely narrow (see arrows). This is due to the fact that the cone of refracted light ($\theta_i < \theta_c$) subtended from the light origin to the crystal bottom plane and exiting the crystal on first pass, and that light cone which reflects

20   off the top and subsequently exits the bottom surface are narrower for closer distances to the photodetector surface. Thus, the base of the detected light cone will also be narrow. The light rays that undergo total internal reflection ($\theta_i > \theta_c$) will for the most part be trapped within the crystal and not contribute significantly to the tails of the distribution.

25

Light distribution shapes were quantified by fitting horizontal profiles through the center of the light distributions shown in FIG. 10B to a Lorentzian function:

5
$$f(x)=Aw/[(x-x_0)^2+(w/2)^2)],$$

where x_0 and w respectively, are the peak location and full- width-at-half-maximum (FWHM) of the distribution $f(x)$. The extracted FWHM values and a sample fit (distribution 5 of FIG. 10B) are shown in FIGs. 11A and 11B. The results in FIGs. 11A and 11B indicate that the FWHM of the light spread profiles typically

10 do not extend more than 1 mm laterally, with the broadest spread (1.5 mm FWHM) occurring when the light is created 1 mm (the furthest distance possible) from the photodetector face. In all cases, at least 90% of the light is contained within 3 mm laterally. For example using a line array with 1 mm wide line array pixels, this

15 suggests that only three APD elements may be necessary to collect a high fraction of the available light created within a crystal sheet. An event could be positioned using a weighted mean on only three signals, avoiding excessive uncorrelated noise propagation, or by selecting the pixel with maximum signal. With a weighted mean approach, resolution would be less than the pixel width. Thus, this experiment

20 indicates that there would be nearly optimal light collection. Of course, with a different photodetector, different positioning characteristics would result.

While specific embodiments of the present invention have been shown and described, it should be understood that other modifications, substitutions and alternatives are apparent to one of ordinary skill in the art. Such modifications,

25 substitutions and alternatives can be made without departing from the spirit and scope of the invention, which should be determined from the appended claims.

Various features of the invention are set forth in the appended claims.